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CONTRACT NUMBER: DAMD17-94-C-4127

TITLE: Real-Time 3D Ultrasound for Physiological Monitoring

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Richland, Washington 99352

REPORT DATE: October 1995

TYPE OF REPORT: Annual

PREPARED FOR: U.S. Army Medical Research and Materiel Command
Fort Detrick, Maryland 21702-5012

DISTRIBUTION STATEMENT: Approved for public release;
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REPORT DOCUMENTATION PAGE

Form Approved
OMB No. 0704-0188

1. AGENCY USE ONLY *(Leave blank)*

2. REPORT DATE
October 1995

3. REPORT TYPE AND DATES COVERED
Annual 15 Sep 94 - 14 Sep 95

4. TITLE AND SUBTITLE
Real-Time 3D Ultrasound for Physiological Monitoring

5. FUNDING NUMBERS
DAMD17-94-C-4127

6. AUTHOR(S)
Steven R. Doctor

7. PERFORMING ORGANIZATION NAME(S) AND ADDRESS(ES)
Battelle Pacific Northwest Laboratory
Richland, Wasington 99352

8. PERFORMING ORGANIZATION REPORT NUMBER

9. SPONSORING/MONITORING AGENCY NAME(S) AND ADDRESS(ES)
U.S. Army Medical Research and Materiel Command
Fort Detrick, Maryland 21702-5012

10. SPONSORING/MONITORING AGENCY REPORT NUMBER

11. SUPPLEMENTARY NOTES

12a. DISTRIBUTION/AVAILABILITY STATEMENT
Approved for public release; distribution unlimited

12b. DISTRIBUTION CODE

13. ABSTRACT *(Maximum 200 words)*

This report documents the first year of a 3-year project to develop a field prototype Advanced Imaging System (AIMS). This prototype will consist of a light-weight, portable ultrasonic imaging system containing a two-dimensional transducer array, computer hardware and software for real-time 3-D holographic image reconstruction and visualization, and a stereovision headset for 3-D image display. During the reporting period, technology was developed to acquire and reconstruct 3-D images in the laboratory, using ultrasonic holography with a point transducer and 2-axis mechanical scanning, and to render these images using volume visualization. In addition, a 3-D imaging testbed system based on "sequential B-scan" technology was developed and placed at Madigan Army Medical Center to gain extensive physician experience with in-vivo 3-D ultrasound.

14. SUBJECT TERMS
Ultrasound, medical imaging, holography, segmentation, volume visualization

15. NUMBER OF PAGES
33

16. PRICE CODE

17. SECURITY CLASSIFICATION OF REPORT	18. SECURITY CLASSIFICATION OF THIS PAGE	19. SECURITY CLASSIFICATION OF ABSTRACT	20. LIMITATION OF ABSTRACT
Unclassified	Unclassified	Unclassified	Unlimited

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Richard J Littlefield 10/30/95

PI - Signature Date

for Steven R. Doctor

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1. INTRODUCTION

Inexpensive, portable diagnostic imaging systems can play a key role in decreasing battlefield fatalities and reducing the cost of military health care. Ultrasound imaging is a particularly promising modality because it does not use ionizing radiation (unlike X-rays), does not require large heavy equipment (unlike magnetic resonance imaging), and has been shown through long use to be safe and effective when used by highly trained practitioners.

However, in current practice, ultrasound produces only two-dimensional images, representing slices through the body. These can be quite difficult to interpret, requiring an ultrasonographer to mentally integrate the information to form an image of the suspect area. Considerable skill is often required even to position the sensor so as to acquire a potentially useful image or combination of images.

Three-dimensional (3-D) ultrasound imaging offers the potential to solve these problems, thus providing a diagnostically valuable, low-cost, real-time imaging modality suitable for operation and use under emergency conditions by non-specialists. Because 3-D volumes show more context than 2-D slices, it becomes easier for users to understand spatial relationships and detect abnormal conditions. Positioning the sensor so as to acquire useful images is also easier with 3-D, because volumetric data can readily be rotated and realigned to good viewing positions, largely independent of the original sensor position. This potentially allows useful 3-D ultrasound data to be taken by an inexperienced operator, then transmitted to and interpreted by a remote expert.

While 3-D ultrasound imaging has been investigated periodically for over 20 years [Robinson 1972], its adoption into routine use has been hindered by clumsy equipment, long image acquisition times, and the difficulties of visualizing the 3-D clouds of relatively noisy data produced by speckle and directional effects of ultrasound. However, recent advances in transducer array technology, computational hardware speed, and improved image reconstruction and visualization methods now appear sufficient to permit these obstacles to be overcome.

These considerations prompted Battelle to submit to ARPA, in response to solicitation BAA94-14 in early 1994, a proposal titled "Real-Time High Resolution 3-D Ultrasonic Imaging for Physiological Monitoring". This proposal laid out the vision of a three stage effort, roughly 8 years in length, leading to the development of an imaging "bed", roughly 10,000 square centimeters in size, containing an array of high resolution ultrasonic

transducers and providing real-time 3-D visualization of many physiological and anatomical structures.

The first stage of this vision, and the focus of the proposal, was a 3-year project to develop a field prototype Advanced IMaging System (AIMS). This prototype would consist of a lightweight, portable ultrasonic imaging system containing a 5 cm by 5 cm two-dimensional transducer array, computer hardware and software for real-time 3-D holographic image reconstruction and visualization, and a stereovision headset for 3-D image display. The system was envisioned as being used to rapidly detect foreign objects and bleeding in the body cavity, lungs, or extremities.

The Battelle proposal to develop an AIMS prototype was accepted by ARPA, and the project began in September 1994. The project consists of two major technical efforts, which proceed in parallel and are coordinated so as to produce periodic milestone deliverables as follows:

- Videotape documenting 3-D holographic image reconstruction,using a single point transducer with 2-D mechanical scanning to simulate an electronically scanned 2-D transducer array. (6/30/95, milestone completed on time and within budget)
- Demonstration of 3-D holographic image reconstruction using an electronically scanned 1-D (linear) transducer array with mechanical scanning on the second axis. (Scheduled for 3/31/96, but likely to slip due to reordering of program activities and funding discontinuities.)
- Demonstration of fully functional brassboard system with 2-D array transducer but without real-time processing speed. (Scheduled for 5/31/97.)
- Demonstration of portable field prototype, including 2-D array transducer with real-time holographic data processing and display. (Scheduled for 6/30/98.)

One major technical effort is to develop transducer, electronic, and computer technology to acquire and reconstruct 3-D ultrasound image using wideband impulse holography [Sheen et.al. 1996]. In overview, this technique works by emitting unfocused (spatially diverging) ultrasound pulses, electronically digitizing the waveforms of the reflected echos, and computationally reconstructing the 3-D image using a frequency domain synthetic aperture approach. Compared with other ultrasonic imaging techniques, such as conventional pulse echo using focused beams, wideband impulse holography has the

unique advantage of providing theoretically optimal resolution and sharp focus throughout the imaged volume. This development of transducer, electronic, and computer technology for holographic imaging is approached mainly through laboratory work, using ultrasound phantoms for evaluation.

The other major technical effort is to develop computer graphic techniques for 3-D volumetric segmentation, visualization, and user interface, with the goal of providing the end-user (physician, medic, paramedic, etc.) with diagnostically valuable information in an easily controlled and interpreted format. Ultrasound data is more difficult to segment and visualize than other 3-D imaging modalities such as CT and MRI, and overcoming these obstacles will be critical to the success of 3-D ultrasound. This primarily software development effort is approached as a collaboration between computer scientists at Battelle and clinical physicians at Madigan Army Medical Center. Software developed at Battelle is tested and evaluated by physicians, using in-vivo 3-D ultrasound data collected at Madigan using a testbed "sequential B-scan" system developed as part of the AIMS project.

In addition to the major technical thrusts, a significant effort is spent to ensure proper targeting and eventual delivery of AIMS technology. This is done through interaction with the medical community and ultrasound companies, and focuses on such issues as choice of frequency & resolution, design and availability of acceptable medical ultrasound phantoms, and manufacturability of transducer arrays.

Excellent progress has been made in all of these efforts during FY95. Details are documented in the following section of this report and in the two technical papers provided as appendices.

2. EXPERIMENTAL METHODS AND RESULTS

As described above, the AIMS project has two major technical efforts:

1. development of 3-D ultrasonic wideband impulse holography; and
2. development of 3-D segmentation, visualization, and user interface software.

Experimental methods and results of the first effort, through June 1995, are described by Sheen, Collins, and Gribble [Sheen et.al. 1996] in a technical publication provided as Appendix 1.

Experimental methods and results of the second effort, through June 1995, are described by Littlefield, Heiland, and Macedonia [Littlefield et.al. 1996] in a technical publication provided as Appendix 2.

Following the June 1995 delivery of the first demonstration videotape, the project team did a significant amount of replanning, resulting in changes to some details of the project plan. There were two major reasons for the replanning, as follows.

1. Interaction with Madigan Army Medical Center personnel indicated a need for earlier and more extensive involvement by representative users (e.g., physicians) than originally had been planned. This is because the medical community has little experience in using 3-D ultrasound, so there is significant uncertainty about what kind of visualization and user interface techniques would be effective for diagnostic purposes.
2. The project's holographic imaging studies prior to June had been done using commercial medical ultrasound plastic phantoms (breast and liver) designed for use at 3.5 MHz and above. Holographic results at 1 MHz were disappointing due to insufficient backscatter in the bulk material. Excellent holographic images were obtained at 5 MHz, but a subsequent analysis of the computational requirements and anticipated 2-D array fabrication technology indicated that 5 MHz was too high a frequency to meet schedule and budget constraints for the current AIMS project. This required the project to become more tightly focused on medical conditions that are both life-threatening and amenable to low-frequency imaging.

In response to these considerations, two notable changes in the project plan were discussed with Dr. Rick Satava, ARPA program manager for the AIMS project. Both changes were approved for implementation.

The first notable change is that the project team decided to design and build a testbed 3-D ultrasound system using "sequential B-scan" technology, and to place this unit at Madigan where it could be used for in-vivo scanning on a routine basis. (The new testbed system was developed in collaboration with Dr. Macedonia and was based on technology developed in his earlier work, refined and packaged to be used routinely instead of only by special effort and scheduling of shared equipment.) The new system was completed and placed at Madigan in late September 1995.

This testbed system utilizes a commercial pulse-echo B-scan ultrasound scanner (Hitachi EUB-405PLUS), combined with a custom-designed lightweight battery powered "paddle"

that mechanically moves the Hitachi transducer probe under precision control. In this testbed system, the Hitachi scanner generates a conventional two-dimensional image, which is digitized into an Apple PowerPC 8100/100 system using a Scion VG5 card. By moving the probe, a sequence of two-dimensional slices is obtained, which together comprise a 3-D ultrasound dataset. A modified version of the NIH Image program [Rasband 1992], extended to include the visualization software developed earlier in the AIMS project, is then used to render 3-D images. Either individual images or an animated sequence of images can be displayed on the computer screen or in stereo on a "Virtual i-glasses!" headset [Virtual i-O, 1995].

The primary purpose of this unit is to be used by Dr. Christian Macedonia and other Madigan physicians in IRB-approved in-vivo studies, to gather extensive sets of 3-D ultrasound images for use in developing and evaluating visualization and user interface techniques. An added benefit is that the unit is suitable for use as a system concept demonstrator. In fact, this system (with a PowerPC 7100 replacing the 8100) was demonstrated by Dr. Macedonia in a man-portable backpack configuration at the October 1995 AUSA meeting in Washington DC. Observers of these demonstrations showed keen interest in the technology, and the project team expects this exposure to result in significant additional contacts regarding requirements and potential uses of AIMS technology.

The second notable change is that the project's holography development effort was refocused much more specifically on the single medical problem of diagnosing hidden bleeding in the abdomen, and on doing the development primarily with phantoms, using only limited in-vivo testing. This approach derives from discussions held with ultrasonographers at the University of Washington (Dr. Stephen Carter), the University of Texas at Houston Medical School (Dr. Karen Ophir and Dr. Jonathon Ophir), and with a manufacturer of medical ultrasound phantoms (Bill Clayman of ATS Labs). In brief, these discussions indicated a general feeling that even a low frequency (1-2 MHz) imaging system would be useful, and that several techniques are known for constructing abdominal bleeding phantoms that faithfully represent tissue at low frequencies.

As of this writing (October 1995), a detailed replan based on this approach is currently being performed.

3. CONCLUSIONS

During the first year of the AIMS project, experimental results clearly demonstrated the feasibility of using ultrasonic wideband impulse holography to reconstruct high resolution

and sharply focused 3-D images of medical phantoms. This demonstration was accomplished using a point transducer with 2-axis mechanical scan and offline image reconstruction. Much more work remains to be done to develop a 2-D transducer array, supporting electronics, and online real-time reconstruction.

AIMS results regarding 3-D ultrasound visualization suggest that volume rendering approaches are more useful with ultrasound data than are the more conventional surface extraction and surface rendering approaches that are commonly used with CT and MRI data. These results are still tentative. Much more testing and evaluation is required, particularly by physicians and other potential end-users of AIMS technology.

No significant work has yet been performed regarding algorithms for segmentation (e.g., tissue or pooled blood identification and highlighting) of 3-D ultrasound data. Algorithmically assisted segmentation may be very important for non-specialist users operating under emergency conditions, and algorithms to accomplish it should be investigated later in the AIMS project.

Successful adoption of 3-D ultrasound as a routine diagnostic tool will depend both on the basic technology and on how well systems are targeted to the needs of the medical providers who use them. Testbed systems such as the one placed at Madigan Army Medical Center permit physicians to experiment with 3-D ultrasound on a regular basis. Lessons learned from this experience will be invaluable in subsequent development of AIMS system specifications.

4. REFERENCES

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- Sheen, DM, Collins, HD, and Gribble, RP, *"Wideband Holographic 3-D Ultrasonic Imaging of Breast and Liver Phantoms"*, conference proceedings of Medicine Meets Virtual Reality: 4, San Diego, CA, Jan.17-20, 1996. See Appendix 1.

- Virtual i-O Inc., Seattle WA, "Virtual i-glasses!" product, info available through URL http://www.vio.com, 1995.

Appendix 1:

"Virtual Reality Volumetric Display Techniques for 3-D Medical Ultrasound" -- MMVR:4 paper

Virtual Reality Volumetric Display Techniques for 3-D Medical Ultrasound

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Randy W. Heiland, Battelle Northwest (rw_heiland@pnl.gov)
Dr. Christian Macedonia, Madigan Army Medical Center

To be presented at Medicine Meets Virtual Reality: 4 (MMVR:4), San Diego, CA, Jan.17-20, 1996 (paper #36).

Abstract

Ultrasound imaging offers a safe, inexpensive method for obtaining medical data. It is also desirable in that data can be acquired at real-time rates and the necessary hardware can be compact and portable. The work presented here documents our attempts at providing interactive 3-D visualization of ultrasound data. We have found two volume rendering visualization packages to be quite useful and have extended one to perform stereographic volume visualization. Using a relatively inexpensive pair of commercial stereo glasses, we believe we have a found a combination of tools that offers a viable system for enhancing 3-D ultrasound visualization.

Introduction

Ultrasound is the imaging modality of choice in many medical situations because it provides very useful information for diagnosis and is safe and inexpensive. In current practice, ultrasound produces only two-dimensional images, representing slices through the body. These can be quite difficult to interpret, and require the ultrasonographer to mentally integrate the information to form an image of the suspect area. Considerable skill is often required even to position the sensor so as to acquire a potentially useful image or combination of images.

If these difficulties could be addressed, ultrasound could provide a diagnostically valuable, low-cost imaging modality suitable for situations where a rapid diagnosis at a remote site is needed. Currently, the Army (through the ARPA Advanced Biomedical Technologies Initiative) is looking to develop a portable 3-D ultrasonic imaging system for use in diagnosing battlefield injuries. Our work is sponsored by that initiative.

Three-dimensional (3-D) ultrasound offers the potential to alleviate the problems addressed above. Because 3-D volumes show more context than 2-D slices, it becomes easier for users to understand spatial relationships and detect abnormal conditions. Positioning the sensor so as to acquire useful images is also easier with 3-D, because volumetric data can readily be rotated and realigned to good viewing positions, largely independent of the original sensor position.

In order to make 3-D ultrasound a practical reality, however, improvements are required in two areas. The first area requiring improvement is 3-D ultrasound image acquisition. One approach is to mechanically scan the sensor head of a conventional ultrasound unit, perpendicular to its imaging plane, so as to produce a sequence of conventional B-scan images. The sequence is then treated as a 3-D image. This approach is reasonably easy and effective, but suffers from long image acquisition time

(seconds) and from relatively poor focus along one axis due to the thickness of the ultrasound beam. A second approach is to use 2-D arrays of unfocused transducers, combined with holographic or synthetic aperture reconstruction techniques. This approach potentially offers both rapid image acquisition and excellent resolution on all axes, but requires a high computational rate in order to run in real time. At this time, ultrasonic holographic array technology is only at the proof-of-principle stage. Both approaches are being pursued by an ARPA-funded research project currently underway as a collaboration between Battelle Northwest and Madigan Army Medical Center.

The second area requiring improvement is 3-D volumetric display. Ultrasound presents some difficulties that are not found in other 3-D imaging modalities such as CT and MRI. Foremost among these are that 1) ultrasound images are typically quite "cluttered", with significant backscatter and intensity variations occurring throughout tissue volumes, and that 2) because of image aperture limitations and because ultrasound is inherently directional, the apparent brightness of tissues and interfaces depends on their position and orientation. Combined, these two aspects make it extremely difficult to explicitly reconstruct anatomically correct 3-D surfaces from ultrasound images. This is a sharp contrast to CT and MRI, where explicit surface reconstruction and display is often the method of choice. Thus, in dealing with 3-D ultrasound, we are faced with the dilemma that humans are not good at visualizing 3-D "clouds" of data, but that is exactly what the imaging process provides. An excellent article that presents the state of the art circa 1992 is found in [Bajura]. Our work aims to address two open-ended problems discussed in that work: visual cues and real-time volume visualization.

We have had good success in producing easily interpreted displays of 3-D ultrasound data through a unique combination of individually standard display techniques, including:

- stereo display (using a commercial LCD headset),
- real-time rotation and rocking,
- nonlinear intensity and opacity maps,
- a gradient-based lighting model, and
- simple shadows,

all operating on the original 3-D volume (no explicit surface reconstruction).

The remainder of the paper is outlined as follows: in the next section, we provide a general introduction to volume visualization and discuss existing algorithms for visualizing 3-D medical data. Section 3 describes our experiences using volume rendering to visualize 3-D ultrasound data. The last section summarizes and discusses plans for future related work.

Volume Visualization

Volume visualization can be defined as a process that uses computer graphics to display structure contained in a three-dimensional dataset. This volumetric dataset can be thought of as a discretized function, F, over some bounded domain. The domain's topology varies from one application to another, but quite often is rectilinear:

$$F(Xi,Yj,Zk),\ i=1,Nx,\ j=1,Ny,\ k=1,Nz$$

This is a generalization of the "cuberille" model [Chen] where the spacings, dx,dy,dz, along the three orthogonal axes are equal. CT and MR data typically have equal spacing along intra-slice directions

(dx=dy), but the distance (dz) between slices is often larger in order to minimize ionizing radiation exposure. Scanned ultrasound data often has the same format. The remainder of this section will be in the context of rectilinear volumetric scalar data.

There are currently two general techniques for visualizing volumetric data: surface rendering and volume or direct rendering. They are inherently different in both the underlying algorithms and the results produced. We provide an overview of the more popular algorithms in the remainder of this section. The distinction in the results produced by the two techniques is that a surface renderer typically generates polygonal geometry, whereas a volume renderer generates an image. Because surface rendering preceded volume rendering in the genealogy of computer graphics algorithms, graphics workstations have evolved to be extremely fast polygon renderers. This, at least in part, accounts for the large collection of papers and software aimed at surface rendering. Volume rendering has become increasingly popular since the SIGGRAPH '88 publication of three papers on the subject [Sabella,Upson,Drebin]. There seems to be an ongoing debate as to whether surface rendering or volume rendering is superior. Little information is published that tries to make a comparison between the two [Udupa]. As we hinted in the Introduction, certain medical imaging modalities may lend themselves better to one visualization technique over another.

Probably the most prevalent algorithm to perform surface rendering on volumetric data is Marching Cubes [Lorensen]. This algorithm constructs a polygonal isosurface from rectilinear scalar data as follows:

- for each cube (voxel) consisting of eight vertices, determine the topology of the isosurface through that cube;
- using an enumerated case table of topologies, compute the geometry of the triangles for the isosurface via some interpolant;
- for improved shading, compute the gradients at the cube vertices and then interpolate to obtain triangle vertex normals.

The SIGGRAPH '88 papers mentioned above offer a glimpse into the classification of volume rendering algorithms. The two broad classes are image-order and object-order. An image-order algorithm casts rays from image pixels into the volumetric data. For each voxel that a ray intersects, relevant information is accumulated to compose the final pixel value. An object-order algorithm operates on the volumetric data in memory-order and essentially projects ("splats") the voxels onto the image in either front-to-back or back-to-front order. The front-to-back method is essentially a volumetric Z-buffer algorithm. What these two classes of algorithms have in common is the basic functionality of mapping the scalar volumetric data, F, into both an intensity (color) function, I(F(x,y,z)), and an opacity function, O(F(x,y,z)). These are collectively known as transfer functions. Allowing a user to (interactively) modify these functions permits improved visualization of relevant structure(s) within the data.

Drebin et al presented an algorithm for transforming the shaded volumetric data into the viewing coordinate system to allow for a projection, via compositing [Porter], along an axis of the stored volume. This work was clarified and extended in [Hanrahan] which decomposed the viewing transformation into a sequence of three shearing matrices. A further improvement to this basic concept of aligning voxels (object space) with pixels (image space) is described in [Lacroute94]. In this work, object space is mapped into a sheared object space which is ideally suited for projecting (compositing) into image space. Furthermore, the authors take advantage of spatial coherence by implementing scanline-based data structures in both object space and image space in order to speed up the rendering algorithm.

Moreover, they offer three variants of the algorithm:

- render a fixed, preclassified volume, but with an arbitrary viewing transform and arbitrary shading parameters (fastest) {A classified volume is one that has an opacity assigned to each voxel}
- render a fixed volume with an associated min-max octree; arbitrary classification, viewing transform, and shading (slower)
- render a fixed volume with no precomputed data structures (slowest).

An implementation of this algorithm is available via the Internet and is known as the VolPack library.

Another volume-rendering technique uses 3-D texture hardware to speed up the projection of object space to image space [Cabral,Cullip]. One such architecture capable of performing this is the Silicon Graphics RealityEngine. SGI provides an implementation of such an algorithm, known as Volren.

Results

We began our investigation with the visualization of ultrasound data from a breast phantom containing internal cysts. The data was obtained by another research group at Battelle Northwest using a holographic ultrasonic imaging technique [Sheen]. We were initially given a 512x512x128 dataset of floating point values and told that the data should be log scaled for better viewing. To make the data volume more manageable we first averaged each of the 128 slices to 256x256 resolution.

We wanted to first render this particular dataset using a volume rendering technique and therefore downloaded the Volren package. After converting our data into TIFF images, one of the required formats for Volren, we discovered that our machine had only enough texture memory to handle a 128x128x64 (or equivalent size) volume. We therefore averaged our 128 slices down to 128x128 resolution and then averaged pairs of slices to obtain a 128x128x64 volume. Figure 1 shows the resulting volume rendering as well as the Volren graphical user interface (GUI). We found Volren to be a very powerful tool for demonstrating the concept of volume rendering to others. The GUI allowed for interactive control of transfer functions that we did not expect to be possible at the beginning of our project. Combined with dynamic viewing (rotation and scaling), Volren provided us terrific initial results. However, with the demand for the SGI Onyx RealityEngine at a premium, we sought other avenues for visualizing our volumetric data.

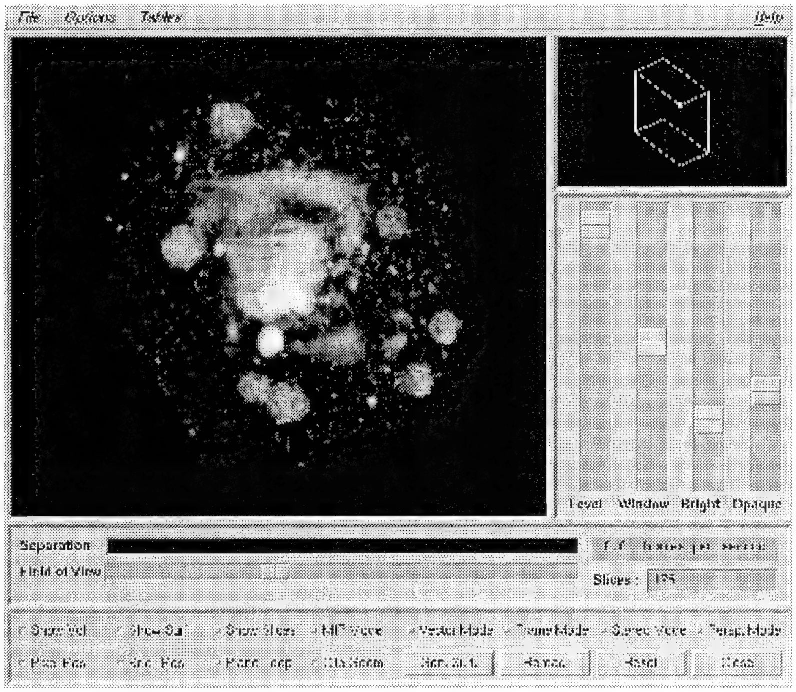

Figure 1. Breast phantom data rendered using Volren.

To satisfy our curiosity at how well a surface renderer would perform on this same data, we computed several isosurfaces of differing isovalues. Figure 2 shows one such surface that closely resembles the Volren rendering. (The isosurfaces were computed and rendered using the AVS visualization package.)

Figure 2. Isosurface of breast phantom data.

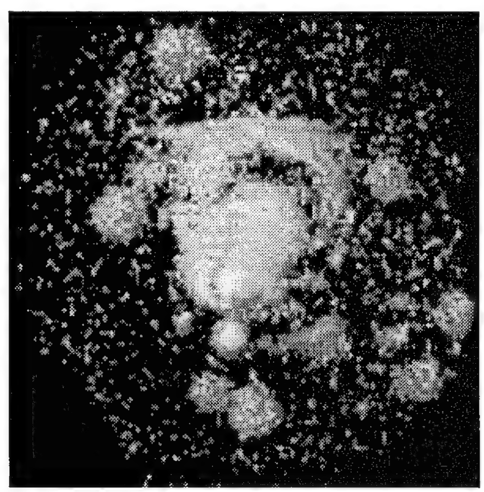

Figure 3. Volume rendering using VolPack.

Next, we downloaded the VolPack volume rendering library and began learning the mechanics of it. There is sufficient documentation in the form of a user's guide, man pages, and example datasets and programs. Though we were no longer bound by the texture memory limit, we kept the breast phantom data at 128x128x64 for comparison. Figure 3 shows an image rendered using VolPack with appropriately chosen transfer functions to approximate the Volren and isosurface renderings.

Figure 4 is the first frame in an MPEG animation showing a rotation of the breast data. These images were generated using the VolPack library. Note that depth-cueing is a user-specified option.

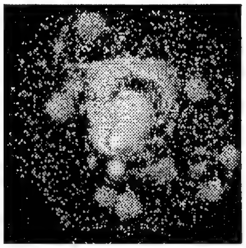

Figure 4. MPEG rotation (164K, 128x128)

In Figure 5, we show a volume rendered image, using Volren, of the face of a fetus with a cleft palate. The data was taken with a conventional B-scan ultrasound imaging unit. The original volume consists of 50 slices at 190x236 resolution. Figure 6 is the first image of an MPEG animation generated using VolPack. Note the shadows which, in addition to the lighting model, serve as helpful 3-D visual cues. Here, we have chosen transfer functions that allow for deeper penetration of the facial skin in an attempt to visually reveal the cleft palate.

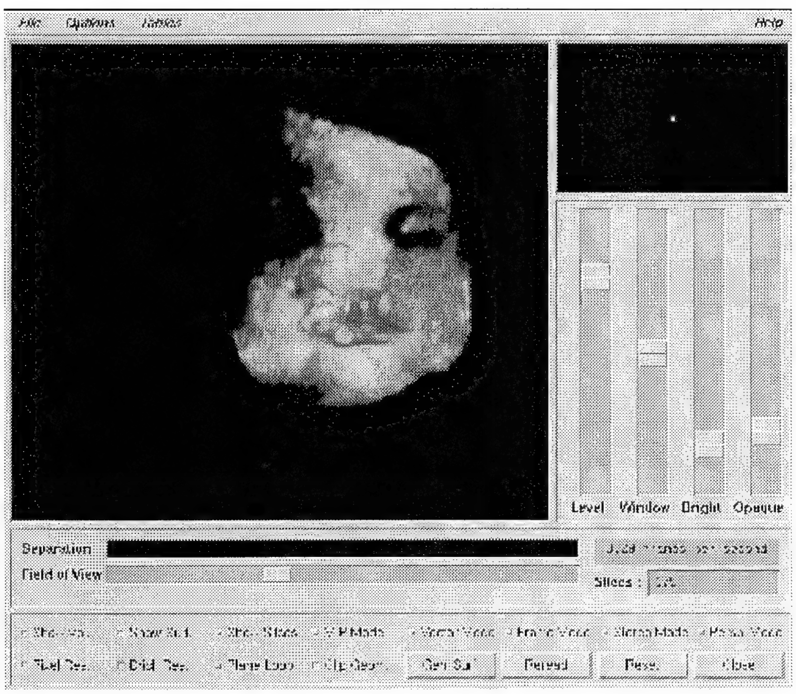

Figure 5. Volren on fetus face.

Figure 6. MPEG animation (272K, 256x256)

Figure 7 shows the image from a volume rendering of another fetus dataset. Again the related MPEG animation was generated using VolPack and demonstrates its scaling and translation transformations.

Figure 7. MPEG animation (107K, 256x256)

One of our goals for this project was to provide stereoscopic displays of volume rendered images. Using the VolPack library, we generated left and right stereo pairs of images. From these we could provide either cross-eye stereo pairs or interlaced scanline stereo images which we viewed using the Virtual i/o i-glasses! (TM) shown in Figure 8.

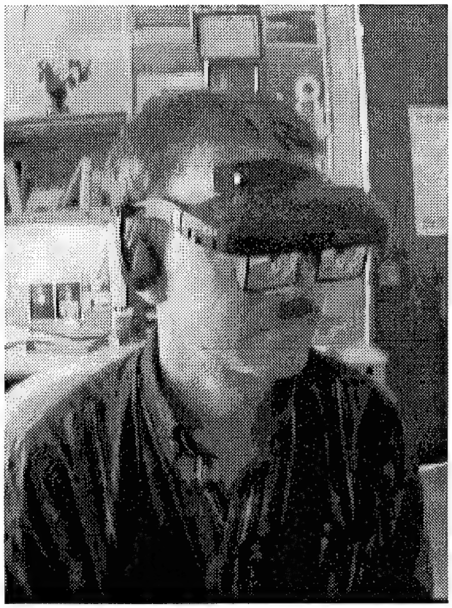

Figure 8. The Virtual i.o i-glasses!

Figures 9 and 10 are MPEG rotations (fast and slow) of cross-eye stereo images. Figure 11 is an MPEG rotation of interlaced scanline stereo images which can be properly viewed wearing a pair of i-glasses! (TM).

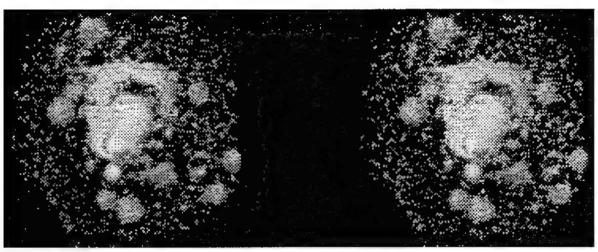

Figure 9. MPEG rotation (132K, 320x128)

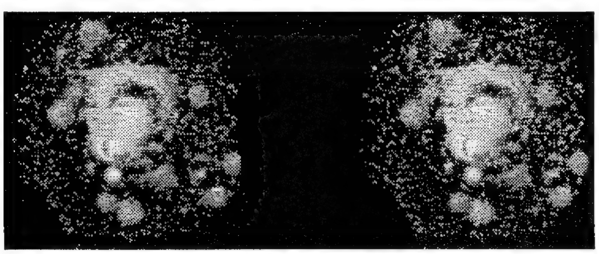

Figure 10. MPEG rotation (486K, 320x128)

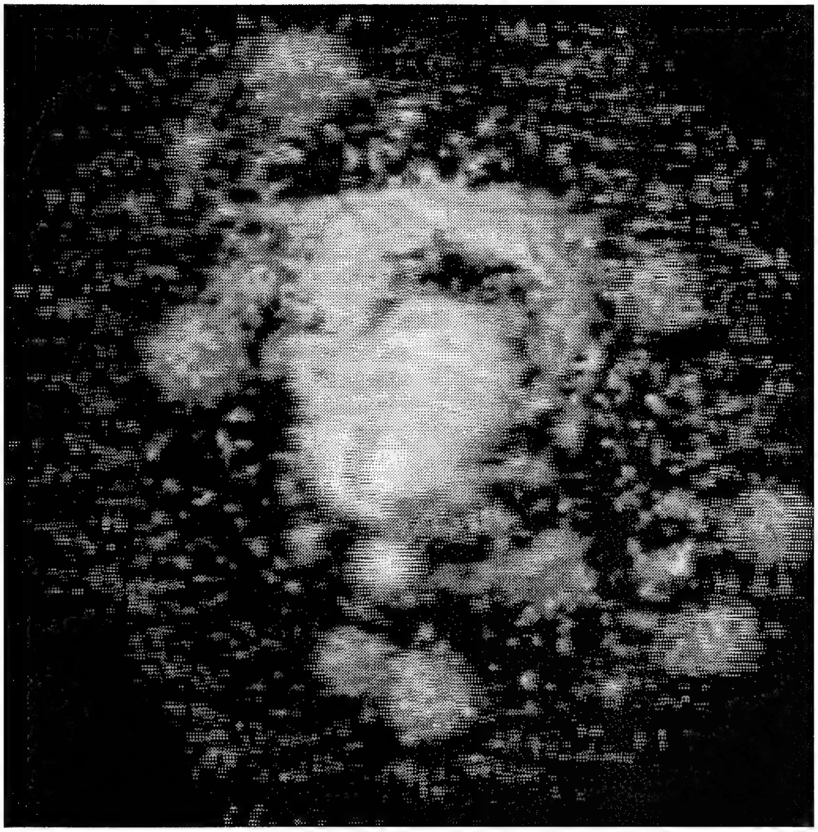

Figure 11. MPEG rotation (943K, 440x440)

In addition to visualizing ultrasound data, we also experimented with volume-rendering CT and MR data from the Visible Human Project. Figure 12 shows a slice of CT data from the lower abdomen with clipping lines that we used to delimit a subregion around the spine. Extracting a volume of data in this fashion, we used the VolPack library to generate the images in Figures 13 and 14. Here we have chosen appropriate transfer functions to highlight the spine (bone) from the rest of the tissue.

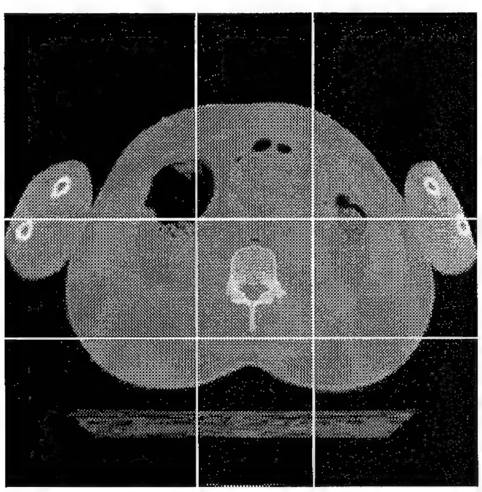

Figure 12. CT slice with clipping lines.

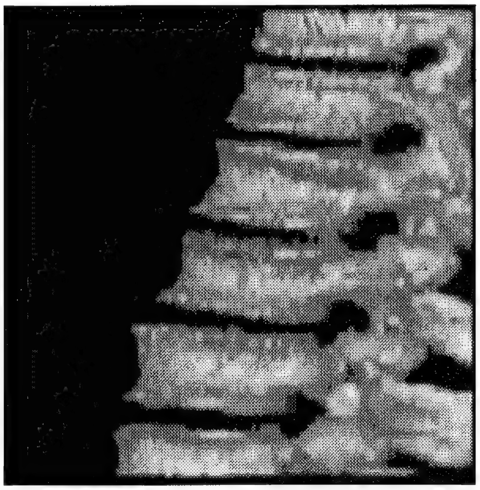

Figure 13. Lower spine from CT data.

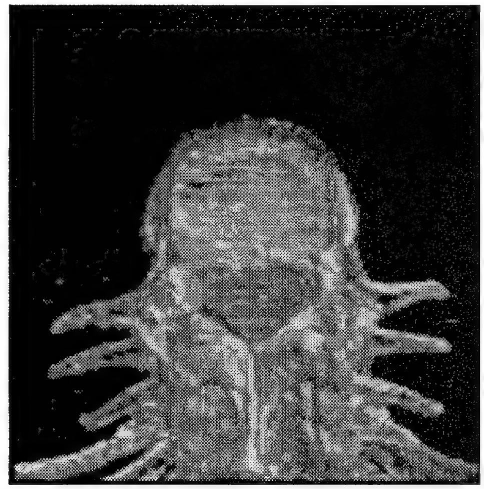

Figure 14. Another view.

Future Work

In a related project at Battelle Northwest, we are investigating extensions to the work described here. We have embedded the stereographic volume-rendering into an immersive virtual environment that allows a user to interactively view and perform simple editing of the volumetric data [Heiland]. We are in the process of porting the VolPack library and our stereographic image application to a Macintosh PowerPC as an extension to the NIH Image application. Finally, we plan to incorporate and experiment with a parallel version of VolPack when it becomes available [Lacroute95].

Acknowledgments

This work was supported by funds from the Advanced Research Projects Agency (ARPA), contract number DAMD17-94-C-4127. Initial research in volume visualization was funded by a Laboratory Directed Research and Development (LDRD) project through the Environmental Molecular Sciences Laboratory (EMSL) at Pacific Northwest Laboratory (PNL). Additional support was through a LDRD project in the Medical Technologies and Systems Initiative at PNL. PNL is a multiprogram national laboratory operated by Battelle Memorial Institute for the U.S. Department of Energy under Contract DE-AC06-76RLO 1830. We would like to thank Dr. Rick Satava of ARPA for his consistent encouragement, Phil Lacroute for timely replies to questions about VolPack, Todd Kulick and Brian Cabral for answering questions about Volren, Dale Collins, Dave Sheen, and Parks Gribble for providing us holographic ultrasound data, and the National Library of Medicine and the Visible Human Project for CT and MR data.

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Appendix 2:

"Wideband Holographic 3-D Ultrasonic Imaging of Breast and Liver Phantoms" -- MMVR:4 paper

Wideband Holographic 3-D Ultrasonic Imaging of Breast and Liver Phantoms

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MMVR Paper Number 11

Abstract

Wideband ultrasonic 3-D holography is a unique technique for volumetric imaging with extremely high lateral and depth resolution. The large frequency bandwidth, which is typically 25% to 100%, provides excellent depth resolution. The synthetic aperture provides optimum lateral resolution of one-half wavelength at the center pulse frequency. Wideband impulse holography is a multi-frequency detection and imaging technique where the reflected target's broadband time waveform signals are recorded over a defined aperture. The signals are then decomposed into their discrete frequency components as single frequency holograms, combined in the spatial frequency domain, and reconstructed into a 3-D composite image. The composite image may then be viewed with stereo glasses in 3-D. Recent 3-D holographic images of human female breast and liver phantoms with internal cysts (5 to 10 mm) at 5 MHz illustrate the efficacy of this technique for medical ultrasound 3-D volumetric imaging. A video of the breast and liver phantoms may be presented at the meeting.

1. Introduction

Ultrasonic medical imaging systems presently in use form two-dimensional images of slices (B-scans) through the human body. These systems are very useful, particularly since they are real-time and form many images per second. This allows the operator to quickly scan through volumes in the body looking for the anatomical features of interest.

Three-dimensional ultrasonic imaging offers significant benefits over conventional B-scan imaging systems for medical systems. One benefit is that a complete scan could be obtained quickly from the volume of interest, with analysis performed in detail at a later time. This analysis could include 3-D rendering with rotation, tilt, as well as the display of image slices through the 3-D image. Some of these image slices could be selected at angles that are not possible with conventional B-scanners due to the lack of a suitable acoustic window through the body at that angle. These tools could aid in the analysis of the images and subsequent medical diagnosis. Another benefit is that true 3-D imaging could be high-resolution in all three-axes. B-scan systems are typically high-resolution only in the two axes displayed in the image, with the elevation direction lower resolution due to the height of the transducer array.

Three-dimensional image acquisition requires that ultrasonic data be gathered over two lateral dimensions with the time axis related to the third image axis. This data can be acquired using focused transducers, or using unfocused transducers combined with computer processing to focus the data.

Current real-time B-scan systems use curved or linear arrays of transducers which are time-delayed to form a scanning focused beam. This data is focused on transmit and receive and requires little or no computer processing to form the image. This type of focusing could be extended to three-dimensions by scanning or rotating the B-scan array. This technique would not provide optimal resolution in the direction of the scan since only a fixed focus would be available along that axis. A more sophisticated way to modify the time-delay focused systems to 3-D would be to fabricate a 2-D array of transducers, with time delay focusing along both axes. Such arrays are currently not available with the large numbers of elements required for medical use.

Wideband holography is another technique that may be used to focus ultrasonic image data. This technique gathers the data by using an unfocused (spherically diverging) beam with the image formed entirely in post-processing using computer processing to form the images. This technique can be used to form images of the highest possible resolution, since each point in the three-dimensional image is focused optimally. It is not required to separate the image space into distinct focal zones. Wideband holography is similar to traditional acoustic holography [1,2, and 3], with the extension to full wideband processing for true depth resolution. A three-dimensional image will require that the diverging beam transducer be scanned over a two-dimensional aperture, or that a linear array be scanned over a linear aperture. A 2-D transducer array could be used to form images with no mechanical scanner required. However, suitable arrays are not yet available. For the experimental results presented in Section 4, a single transducer was scanned over 2-D aperture to obtain the 3-D image data.

At this stage, our research has been directed at obtaining 3-D ultrasonic images of the highest resolution using medical imaging phantoms. As presently implemented, our experimental techniques require more time than would be available for practical medical imaging systems. Ongoing efforts, are directed at modifying these techniques to allow for fast data acquisition and processing. This work is part of an ARPA sponsored research project directed at the development of 3-D ultrasonic imaging systems useful for the diagnosis of battlefield injuries.

2. Wideband Holographic Imaging Technique

Wideband holographic imaging is a means of forming high resolution three-dimensional images of targets from wide-bandwidth data gathered over a two-dimensional planar aperture. The target to be imaged is illuminated over a planar aperture using an unfocused, or diverging, coherent source such as an ultrasonic transducer. The transmitted diverging beam impinges on the target resulting in echoes which are recorded by the transducer. The reflected or echoed signal is then recorded coherently (digitized) and stored in a computer memory. The complete set of unfocused data can then be used to mathematically reconstruct a fully focused three dimensional image of the target's reflectivity function.

A highly efficient algorithm has been developed that may be used to perform this image reconstruction. This technique relies extensively on the use of Fourier Transforms which may be computed very efficiently using the Fast Fourier Transform algorithm. The reconstruction method allows for wide relative bandwidths and allows for the targets to be near to the scanned aperture. No far-field approximations are made. Therefore, the only limitation on the resolution obtained is the diffraction limited resolution imposed by the wavelength, source and receiver beamwidths, size of aperture, and distance to the target. This method is similar to single frequency holography [1, 2, and 3], with the extension to wideband illumination, which allows for true three-dimensional high resolution imagery from a two-dimensional planar aperture.

Data collection is performed by scanning a transmitting source and receiver over a rectilinear planar aperture which has one or more targets within its field of view. The image reconstruction algorithm described in this section is an extension of work by Soumekh [4, 5]. Soumekh derived a wideband synthetic aperture imaging algorithm which reconstructs data from a linear aperture into a two-dimensional image. This work extends this derivation by making the aperture planar instead of linear which allows for a fully three-dimensional image reconstruction.

A full mathematical derivation of the reconstruction algorithm is beyond the scope of this paper, however, an overview of the technique will be given. The transducer is assumed to produce a diverging beam. At a modest distance from the transducer this beam is approximately a spherical wave. This spherical-wave can be mathematically decomposed into a superposition of plane-wave components emanating from the source over all possible angles. The plane-wave decomposition is performed in the frequency domain, so the recorded time-domain data are Fourier Transformed to obtain the frequency domain data. After plane-wave decomposition the spherical wave phase function is expressed in terms of a superposition of linear phase functions. These linear phase functions are plane waves and may be grouped as Fourier integrals and inverted using multi-dimensional Fourier Transforms. Thus, an efficient image reconstruction is possible based almost entirely on the Fourier Transform. For computer implementation, the Fast Fourier Transform algorithm is used resulting in a very efficient image reconstruction algorithm.

The expected lateral resolution for the wideband holographic imaging system is approximately one-half wavelength times the F-number. The F-number is equal to the range to the target divided by the aperture width, or is determined by the beamwidth of the transducer. The longitudinal range resolution is determined by the pulse width or bandwidth of the system, which are inter-related. The range resolution may be expressed as v/2B, where v is the velocity of sound in the medium, and B is the bandwidth of the system. The range resolution may also be expressed as vT/2 where T is the pulse width in time. For a wideband system, both the lateral and range resolutions obtained are on the order of one-half to four wavelengths at the center frequency.

3. Experimental Imaging System

An experimental imaging system was assembled to gather wideband holographic ultrasonic data from various medical ultrasonic phantoms. This system consists of a high-resolution two-axis mechanical scanner, a water bath, an ultrasonic transducer assembly operating at 5 MHz with associated transmit and receive circuitry, and a computer with A/D converter.

The mechanical scanner is high-resolution and can scan up to 15 cm on each of two axes. The scanner holds the transducer a short distance above the target . The water bath allows the ultrasonic waves to be coupled to the target, while still maintaining a planar scanned aperture. An attractive feature of the water-bath system is that the shape of the target is undisturbed and can be imaged along with the underlying features of the target. This is not possible using conventional B-scanning techniques in which the scanner is directly coupled to the target and therefore does not accurately image the shape of the target.

The transducer assembly consists of two small (60 mil) circular flat transducers operating at 5 MHz. At this frequency the transducers are approximately 4 wavelengths wide, and produce a beamwidth of approximately 14 degrees. This beamwidth results in an F-number approximately equal to 4. An F-number of 4 will produce lateral resolution of approximately 2 wavelengths. For the imaging results

presented in the next section, the transmitted waveform consisted of 8 wavelengths at 5 MHz. This results in a depth resolution of approximately 4 wavelengths, or twice the lateral resolution dimension. The transmitted waveform (8 cycles) has a bandwidth of approximately 25%. At 5 MHz the wavelength of ultrasound in water is approximately 0.3 mm. This configuration therefore has a lateral resolution size of approximately 0.6 mm, and a depth resolution size of approximately 1.2 mm. The depth resolution could easily be improved by reducing the number of cycles transmitted. This was not done so that the size of the overall data set would be more manageable.

The receiver circuitry amplifies the target echoes prior to A/D conversion. If digitized directly, the received 5 MHz waveform would require sampling rates in excess of at least 10 MHz and more typically 20 MHz (allowing for 4 samples per wavelength). Since the bandwidth of our system is only approximately 25%, the sampling rate can be reduced considerably by coherently demodulating the received signal. This is done by mixing it in quadrature with the source oscillator (5 MHz) used to generate the 8 transmitted cycles, to yield in-phase(I) and quadrature (Q) demodulated signals. The full-bandwidth of the system is 25% of 5 MHz or 1.25 MHz. Thus, the maximum frequency in each of the I and Q signals is 0.625 MHz. The demodulated I and Q signals must be sampled at a rate of at least 1.25 MHz or more typically 2.5 MHz. Our system was configured with a sample rate of 5 MHz to allow flexibility to vary the system bandwidth without changing the data sampling rate.

Once a complete scan of the target has been digitized and stored in the computer, the computer image reconstruction algorithm discussed in the previous section is applied to the data. The numerical reconstruction is performed using a high-speed arithmetic co-processor. The resulting 3-D image data can then be displayed on the host computer as 2-D slice images, or processed further for 3-D rendered images.

4. Imaging Results

To evaluate the effectiveness of the wideband holographic imaging technique on targets as realistic as possible, ultrasound phantoms of the human female breast and human liver were selected. The breast phantom is of a type used to train physicians for ultrasound guided biopsies. It contains simulated cysts which are in the form of 10 mm diameter spheres. The liver phantom is constructed in a similar fashion to the breast phantom. A simulated gall bladder with gall stones is included in the phantom as well as simulated cysts.

Unlike conventional B-scanners, the output of the wideband holographic imaging system is a fully-focused 3-D image. In this paper, we will however, not attempt to render the data in a three-dimensional fashion but will simply examine various slices through the reconstructed 3-D images. Two-different orthogonal image slices will be shown. B-scan slices are slices through the data set which are perpendicular to the plane containing the scanned aperture. C-scan slices are slices through the data set which are parallel to the scanned aperture. It should be pointed out that the B-scan view is similar to that from conventional B-scanners, it is somewhat different since it will exclude objects outside the relative narrow lateral resolution cell size.

The data presented in this paper has also been rendered in a 3-D manner and will be presented in a separate paper at this conference by R. Littlefield, R. Heiland, and C. Macedonia.

Figures 1 and 2 show image slices from the reconstructed 3-D image of the human breast phantom imaged at 5 MHz. This 3-D image was obtained from an 80 mm by 80 mm by 27.5 mm scan of breast

phantom placed in a water bath. The phantom was scanned from above, i. e. the time axis corresponds to the vertical image axis in Figure 1. To avoid aliasing a large number of data samples were taken with 512 sample points in both lateral dimensions and 256 samples (at 5 MHz sampling) in time. The first image in Figure 1 shows the a cross-section slice of the breast phantom which contains only background scatterers. The second image clearly shows a 10 mm spherical simulated cyst. There is also some shadowing below the cyst. The third image shows another cyst near the bottom right. The fourth image clearly shows the outline of the breast along with two cysts. The nipple and the area just to the right of the nipple are relatively bright in this image. This is due to the normal to the surface of the breast being aligned with the transducer which results in a specular reflection at these points. This type of feature would not be present if the transducer were directly coupled to the target, as is commonly done with conventional imaging. The fifth image shows a bright cyst (echogenic) as well as a darker cyst (non-echogenic).

Figure 2 shows C-scan slices from the reconstructed 3-D image of the human breast phantom. These images clearly resolve the various simulated cysts. Some of the darker circular areas are shadows caused by bright cysts blocking some of the ultrasonic energy, while others are due to non-echogenic cysts. The high lateral resolution is demonstrated by the bright point scatterers located on the surface of the breast.

Figure 3 shows C-scan image slices from the reconstructed 3-D image of the human liver phantom imaged at 5 MHz. This 3-D image was obtained from a 75 mm by 100 mm by 60.5 mm scan of liver phantom placed in a water bath. To avoid aliasing a large number of data samples were taken with 512 sample points in both lateral dimensions and 256 samples (at 5 MHz sampling) in time. The second image in Figure 3 clearly shows a non-echogenic (dark) spherical cyst. The third, fourth, and fifth images clearly show the simulated gall bladder, along with several simulated gall stones. As in Figure 2, the high lateral resolution is demonstrated by the bright point scatterers seen in several of the images in Figure 3.

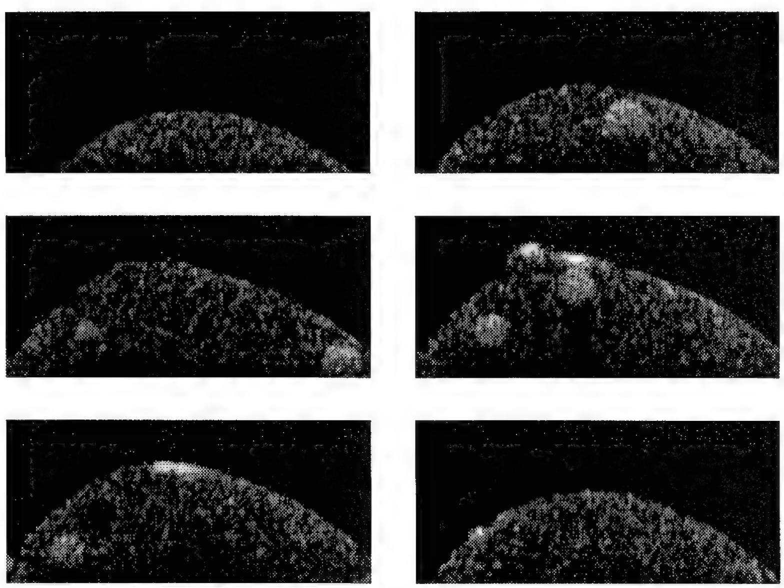


Figure 1: 3-D ultrasonic images of human breast phantom at 5 MHz (B-scan view shown).

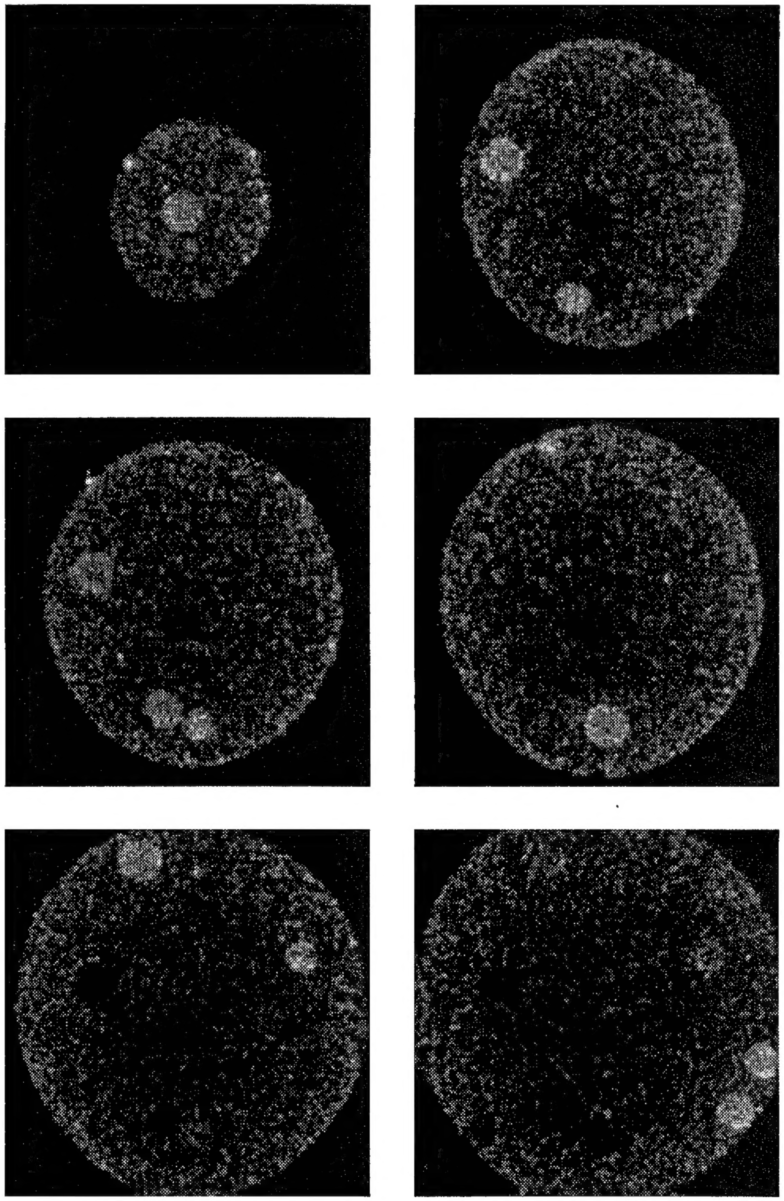
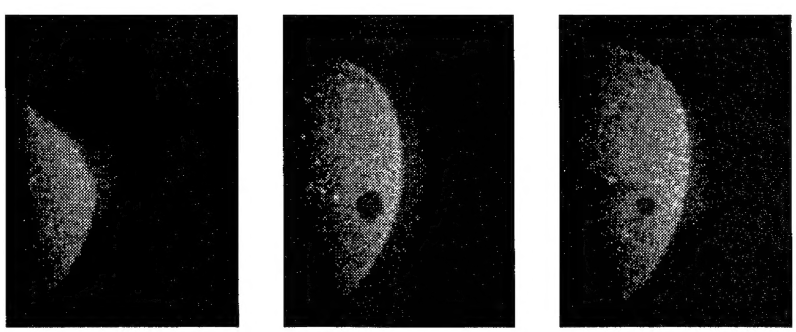


Figure 2: 3-D ultrasonic images of human breast phantom at 5 MHz (C-scan view shown).



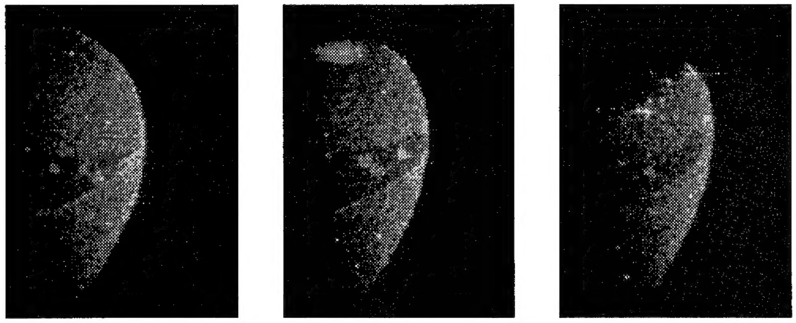


Figure 3: 3-D ultrasonic images of human liver phantom at 5 MHz (C-scan view shown).

5. Conclusions

High resolution three-dimensional ultrasonic imagery of breast and liver phantoms has been demonstrated. The wideband holographic imaging technique has been shown to provide high-resolution simultaneously in both lateral dimensions as well as the range dimension. This high resolution is obtained for all positions throughout the image. This is an improvement over conventional systems which use one or more fixed focal zones to obtain the focused image. In addition, conventional B-scanners will have rather poor resolution in the elevation direction, i.e. the direction normal to the image plane. Difficulties, however, currently prevent the application of this type of technique for clinical use. At high frequencies (such as 5 MHz) the data set is very large and cannot be reconstructed quickly with low-cost computer systems. Furthermore, the velocity of sound limits the number of pulses that may be transmitted per second which limits the frame rate of a 3-D imaging system. Despite these limitations, we believe that synthetic focusing techniques such as the wideband holographic technique, or variations, will play a role in the development of 3-D clinical ultrasonic imaging systems.

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Appendix 3:

June 1995 Deliverable Videotape Contents

Battelle / Pacific Northwest Laboratories
Richland, WA

BNW Project 22258
USAMRAA No. DAMD 17-94-C-4127

Real Time Ultrasound
For Physiological Monitoring

Dr. Steven R. Doctor, Project Manager

DRAFT Progress Report Videotape, June 1995

This videotape is provided as a DRAFT report. It has not been cleared for public release by Battelle/PNL and/or by the Advanced Research Projects Agency.

This tape presents research results in ultrasonic holography and volume visualization.

It contains several sequences:

1. Holographic image of a breast phantom, showing high resolution on all axes, and illustrating an assortment of real-time display and interaction techniques.

2. Side-by-side comparison of holographic image and sequential B-scan image.

3. Volume rendered animations of the breast phantom (ultrasonic holography), human spine (CT), fetal skeleton and fetal face (both sequential B-scan).

4. Same animations, rendered in stereo for display on Virtual I/O "i-glasses!" headset.

Holographic Image of Breast Phantom

5 MHz, single transducer, 2-D mechanical scan, water bath.

Scanned and reconstructed at 512 x 512 x 256, averaged to 128 x 128 x 64 for display purposes.

Spherical targets ("cysts") are 10 mm diameter.

The sequence begins with display parameters chosen to highlight bright "cysts". The image is rotated interactively, and clipping planes are used to expose its interior. Near the end of the sequence, display parameters are reset to show a wider range of echogenicity, including shadows and dark "cysts".

Display platform: Silicon Graphics Reality Engine 2, real-time "volren" software.

Side-by-Side Comparison of Holographic and Sequential B-scan Images

The breast phantom and holographic image are as described for the previous sequence.

The sequential B-scan image was constructed by digitizing a videotaped sequence of scans provided by Dr. Christian Macedonia at Madigan Army Medical Center, using a water bath and a 5 MHz transducer. In the displays shown here, the images have been rotated so that the mechanical scan axis is vertical, i.e., each B-scan is horizontal and the display shows a slice through the middle of 128 B-scans.

The holographic and sequential B-scan images have been aligned to the greatest extent practical, but do not correspond perfectly -- the sequential B-scan image includes a slightly larger part of the phantom.

Volume-Rendered Animation (non-stereo)

The following sequences illustrate the results of a more sophisticated volume rendering algorithm that incorporates a gradient-based lighting model, simple shadows, and improved opacity handling.

Data sources and imaging modalities are:
1. breast phantom: Battelle ultrasonic holography.
2. spine: Visible Human Project, CT.
3. fetal skeleton and fetal face (cleft palate): Dr. Christian Macedonia, Madigan Army Medical Center, sequential B-scan.

These images are rendered using software based on Stanford's "VolPack".

Volume-Rendered Animation (stereo)

These sequences repeat the previous set, but are rendered so as to display in stereo when viewed through the Virtual I/O "i-glasses! Personal Display System" headset.

They should be viewed with the i-glasses! control switch set to position 3D-1.

Note that these images contain left- and right-eye images packed together as alternating scan lines.

When viewed without the i-glasses! or with the i-glasses! control switch set to STD, the images will not be seen in stereo and may flicker badly.